Diagnostic Ultrasound: Principles and Applications

Daniel F. Leotta, PhD

University of Washington
Applied Physics Laboratory
Center for Industrial and Medical Ultrasound
Ultrasound Imaging

Medical imaging modality based on high-frequency sound waves
Ultrasound Imaging

- Ultrasound imaging is based on echo-ranging principles
  - Short-duration sound pulses are transmitted into the body
  - Received echoes are used to construct 2D images of tissue
Ultrasound Instruments

• Key advantages
  • No ionizing radiation
  • Real-time display
  • Anatomy and physiology (Doppler)
  • Relatively inexpensive
  • Portable
Ultrasound Instruments

iU22 (Philips Ultrasound, Bothell)

180PLUS (SonoSite, Bothell) - Late 1990s
Ultrasound Instruments

- GE LogiqBook
- GE Vscan, 2009
Ultrasound Instruments

- Cell phone imager (2009)
- Richard and Zar, Washington U, St. Louis
AFFORDABLE DIAGNOSTICS
AT THE POINT OF CARE

Mobisante, Inc. is a privately-held mHealth company and we are building the world's first SmartPhone-based, Low-cost, Easy-to-use, Portable Ultrasound Imaging Systems. These systems are based on technology developed by a top university research lab and partially funded by Microsoft Research. The basic systems, which are expected to be extremely affordable, will make Ultrasound imaging accessible to primary care physicians, physician extenders, field workers in telemedicine settings, veterinarians and other cost-sensitive medical institutions in rural and emerging markets locally, regionally and around the globe. If you are interested in learning more, please contact us - we are interested in hearing from you.
Ultrasound Imaging

- Limitations
  - Operator dependent
    - Strong angle dependence
  - Complex spatial resolution parameters
  - No record of image locations (in general)
  - Artifacts
Ultrasound Imaging

- Sound is transmitted and received by a piezoelectric transducer.
- Returned echoes vary depending on tissue characteristics.
- Timing is used to determine echo depth.
Sound

Vibration of particles in a medium

– OR –

A mechanical pressure wave

Sound is a longitudinal wave: particle motion is parallel to the direction of propagation

Physics, Cutnell & Johnson, 1992
Sound

Compression and Rarefaction
So, time

Period (T)

Distance (d)

Amplitude

Wavelength (λ)

Frequency = cycles/sec
### Spectrum of Sound

<table>
<thead>
<tr>
<th>Infrasound</th>
<th>Audible Sound</th>
<th>Ultrasound</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>$10^2$</td>
<td>$10^6$</td>
</tr>
<tr>
<td>1</td>
<td>$10^3$</td>
<td>$10^7$</td>
</tr>
<tr>
<td>10</td>
<td>$10^4$</td>
<td>$10^8$</td>
</tr>
<tr>
<td></td>
<td>$10^5$</td>
<td>$10^9$</td>
</tr>
<tr>
<td></td>
<td>$10^6$</td>
<td>$10^{10}$</td>
</tr>
</tbody>
</table>

- **Infrasound**
  - 0 - 10 Hz
  - Music and speech: 20 - 20,000 Hz

- **Audible Sound**
  - 0 - 10 Hz
  - Medical imaging: 1 - 100 MHz

- **Ultrasound**
  - 0 - 10 Hz
  - Medical imaging: 1 - 100 MHz
Ultrasound Imaging

Properties of sound propagation that affect the image
Piezoelectric Effect

- Transducers constructed from piezoelectric materials
  - converts electrical energy to mechanical energy
  - converts mechanical energy to electrical energy
  - the material, shape and size of the transducer influence the frequency it can generate
Beam Pattern

- Characteristic beam pattern for single-element circular transducer
- Near field (Fresnel Zone): non-uniform intensity, non-divergent
- Far field (Fraunhofer Zone): uniform intensity, divergent
- Natural focus at transition between zones

\[ r > \lambda \]

\[ \text{Length of near field} = \frac{r^2}{\lambda} = \frac{d^2}{4\lambda} \]

\[ \text{Angle of beam divergence:} \]
\[ \sin \theta = 1.22 \frac{\lambda}{d} \]
\[ \theta = \arcsin \left( 1.22 \frac{\lambda}{d} \right) \]
**Beam Pattern**

- Beam diameter determined by:
  - distance from transducer
  - transducer diameter
  - frequency

- Deeper near field
- Less divergent far field

**Diameter**

- larger diameter

**Frequency**

- low frequency
- high frequency
Properties of Sound

Reflection

- Production of echoes at interfaces of tissues with different physical properties
  - sound waves that do not transmit across an interface are redirected back into the medium from which they originated
Types of Reflection

- **Specular**
  - Structures larger than the wavelength
    - Angle of reflection = Angle of incidence

- **Scattering**
  - Structures smaller than the wavelength
    - Multi-directional
Types of Reflection

Specular reflection

Non-perpendicular Incidence

Scattering
Reflection

Angle dependence
Reflection

Angle dependence

ROTATOR CUFF TEAR
Amount of Reflection

• Depends on the difference in Acoustic Impedance (Z) of the media
  - how much sound is reflected is directly proportional to the impedance mismatch

\[ Z = \rho c \]

\( \rho \) = density = mass/volume  
\( c \) = propagation speed
## Acoustic Impedance ($Z$)

<table>
<thead>
<tr>
<th>Medium</th>
<th>$Z$ (Rayls)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Air</td>
<td>$0.0004 \times 10^6$</td>
</tr>
<tr>
<td>Fat</td>
<td>$1.38 \times 10^6$</td>
</tr>
<tr>
<td>Water</td>
<td>$1.48 \times 10^6$</td>
</tr>
<tr>
<td>Soft tissue</td>
<td>$1.63 \times 10^6$</td>
</tr>
<tr>
<td>Muscle</td>
<td>$1.70 \times 10^6$</td>
</tr>
<tr>
<td>Bone</td>
<td>$7.80 \times 10^6$</td>
</tr>
</tbody>
</table>

Rayl: kg/m$^2$/s
Properties of Sound

Speed of Sound

How fast sound travels in a given medium depends on the structure of the medium

- Density of particles
- Stiffness
  - Bulk Modulus: resistance to compression
# Speed of Sound (c)

<table>
<thead>
<tr>
<th>Medium</th>
<th>m/sec</th>
</tr>
</thead>
<tbody>
<tr>
<td>Air</td>
<td>331</td>
</tr>
<tr>
<td>Fat</td>
<td>1450</td>
</tr>
<tr>
<td>Water</td>
<td>1482</td>
</tr>
<tr>
<td>Soft tissue</td>
<td>1540</td>
</tr>
<tr>
<td>Blood</td>
<td>1570</td>
</tr>
<tr>
<td>Muscle</td>
<td>1585</td>
</tr>
<tr>
<td>Bone</td>
<td>4080</td>
</tr>
<tr>
<td>Steel</td>
<td>5960</td>
</tr>
</tbody>
</table>
Distance Equation

Convert time to distance to create accurate anatomic images

\[ D = c \cdot t \]

c = sound propagation speed (m/s)

t = time (sec)
**Distance Equation**

Example 1

\[ D = c \cdot t \]

\[ c = 1540 \text{m/sec} \]

*How long does it take for sound to travel 1cm into the body?*

\[ 0.01 \text{m} = 1540 \text{m/sec} \times t \]

\[ t = 0.00000065 \text{sec or } 6.5 \mu \text{sec} \]
Distance Equation

Example 2

D = c t

c = 1540 m/sec

How long does it take for the transducer to receive a signal 4 cm deep?

It takes 6.5 µsec for 1 cm of travel.

6.5 µsec x 4 cm x 2 = 52 µsec

Round-trip travel
Properties of Sound

Attenuation

The decrease in the intensity of sound as it travels through a medium (loss of energy)

- Absorption
- Reflection

- Proportional to distance and frequency
  - Longer path: increased attenuation
  - Higher frequency: increased attenuation
### Attenuation Coefficients

<table>
<thead>
<tr>
<th>Tissue</th>
<th>Attenuation Coefficient (dB/cm/MHz)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Liver</td>
<td>0.5</td>
</tr>
<tr>
<td>Fat</td>
<td>0.6</td>
</tr>
<tr>
<td>Brain</td>
<td>0.6</td>
</tr>
<tr>
<td>Kidney</td>
<td>0.9</td>
</tr>
<tr>
<td>Muscle</td>
<td>1.0</td>
</tr>
<tr>
<td>Heart</td>
<td>1.1</td>
</tr>
</tbody>
</table>

0.8 dB/cm/MHz average value for soft tissue
Attenuation

• To calculate attenuation, multiply the Attenuation Coefficient by round trip distance and frequency:

• 3.5 MHz sound, 4 cm sound travel
  • attenuation = 0.5 dB/cm/MHz x 4 cm x 3.5 MHz = 7 dB

• 5 MHz sound, 10 cm sound travel
  • attenuation = 0.5 dB/cm/MHz x 10 cm x 5 MHz = 25 dB
Ultrasound Imaging

Converting the sound wave into a 2D image
Ultrasound Imaging

Overview: Instrument

- Transmit short pulse into tissue
- Receive echoes
- Perform amplitude demodulation
- Maintain time record time for depth calculation
- Repeat to create 2D map of echo amplitude
Components of an Ultrasound Scanner

2D gray scale imaging

- Transmitter
Pulse-Wave Parameters

- **time**
- **distance**

- **T**
- **PD**
- **PRP**
- **λ**
- **SPL**

**PRF = #pulses/sec**
- **PRF** = *Pulse Repetition Frequency*
  
- **PRP** = *Pulse Repetition Period*  
  = 1/PRF  
  = time until pulse repeats itself

- **PD** = *Pulse Duration* = time pulse lasts

- **DF** = *Duty Factor* = \( \frac{PD}{PRP} \)  
  = active time / total time

- **SPL** = *Spatial Pulse Length*  
  = length of pulse in space  
  = (cycles/pulse) \( \times \) (\( \lambda \))

**Pulse-Wave Parameters**

**Related to temporal resolution**

**Related to spatial resolution**
Typical pulse times

For 3-cycle 3 MHz pulse: PD = 1 microsecond

For 10 cm depth: PRP = 130 microsecond
Components of an Ultrasound Scanner

2D gray scale imaging

• Master clock for pulse and echo timing
Components of an Ultrasound Scanner

2D gray scale imaging

- Short pulse transmitted
- Echoes received and amplified
Components of an Ultrasound Scanner

2D gray scale imaging

- Echo ‘detection’
Signal Processing

- **Amplitude modulation**
  - Changes in amplitude of received energy provide information about tissue characteristics
    - Low-frequency amplitude variations superimposed on high-frequency signal
  - ‘Detection’ or ‘Demodulation’ extracts the information from the returned echoes

- **Frequency modulation**
  - Changes in frequency of received energy provide information about tissue motion
  - Doppler ultrasound
A short pulse is transmitted and a series of echoes is received.

The ultrasound echo is referred to as a ‘Radio Frequency’ signal because it is in the frequency range of radio waves in the electromagnetic spectrum.

### Electromagnetic Spectrum

<table>
<thead>
<tr>
<th>Region</th>
<th>Frequency</th>
</tr>
</thead>
<tbody>
<tr>
<td>Radio</td>
<td>$10^6 - 10^9$</td>
</tr>
<tr>
<td>Microwave</td>
<td>$3 \times 10^{12}$</td>
</tr>
<tr>
<td>Infrared</td>
<td>$4 \times 10^{14}$</td>
</tr>
<tr>
<td>Visible</td>
<td>$7 \times 10^{14}$</td>
</tr>
<tr>
<td>Ultraviolet</td>
<td>$3 \times 10^{17}$</td>
</tr>
<tr>
<td>X-Ray</td>
<td>$3 \times 10^{19}$</td>
</tr>
<tr>
<td>Gamma Ray</td>
<td>$&gt; 3 \times 10^{19}$</td>
</tr>
</tbody>
</table>

Medical ultrasound frequencies $\approx 2$-20 MHz.
**Amplitude Demodulation**

- Separation of carrier and modulating waveform
  - Rectification
    - Absolute value
  - Envelope detection
    - Low-pass filter
Short pulses are transmitted and the echo times are measured.
- A-Mode scan (Amplitude mode)
Components of an Ultrasound Scanner

2D gray scale imaging

- Beam location saved in scan converter memory
- Echo amplitude mapped to gray scale
- 2D image displayed on screen
B-Mode

- Pulses are transmitted in multiple directions and the echoes are mapped to brightness on a 2D display
  - B-Mode scan (Brightness mode)
B-Mode

- Pulses are transmitted in multiple directions and the echoes are mapped to brightness on a 2D display
  - B-Mode scan (Brightness mode)

**B-Mode Imaging System**

**Transducer array**

**Computer display**
Processing Example: RF to B-Mode

Images of test phantom with Terason portable ultrasound scanner

Bitmap image
RF to B-Mode

Amplitude Demodulation

Single RF line
RF to B-Mode

Amplitude Demodulation

Rectified RF
RF to B-Mode

Amplitude Demodulation

Envelope (A-Mode)
RF to B-Mode

A-Mode to B-Mode
RF to B-Mode

A-Mode to B-Mode
RF to B-Mode

A-Mode to B-Mode

128 Lines
Components of an Ultrasound Scanner

2D gray scale imaging

- **Time-Gain Compensation**: adjust gain as a function of time (depth) to compensate for attenuation
Time-Gain Compensation (TGC)

Compensate for signal attenuation as a function of depth
Time-Gain Compensation

- Average attenuation rate: 0.8 dB/cm/MHz
- Variable TGC allows gain adjustment at different depths
- Multiple slider controls
TGC: correct
TGC: incorrect
Ultrasound Scanheads

• Scanhead construction and operation determine the format and characteristics of the ultrasound scan plane
  • Scanhead design affects resolution
    • Spatial and temporal
  • Range of designs available for specific imaging applications
Ultrasound Scanheads

(a) Rotating mechanical device

(b) Linear array: scans an area the same width as the scanhead

(c) Curved linear array: sweeps a sector

(d) Phased array: variable timing of the excitation across elements steers the beam so that a small transducer sweeps a large area
Mechanical Scanhead

- Single-element transducer is swept across the image plane by a motor
  - Prone to wear and damage over time
  - Fixed focus
Focusing Methods

Curved transducer face

Lens

Electronic focusing (phasing)

Only **electronic** focusing allows for variable focus; all other methods have fixed focus
Ultrasound Scanheads

(a) Rotating mechanical device

(b) Linear array: scans an area the same width as the scanhead

(c) Curved linear array: sweeps a sector

(d) Phased array: variable timing of the excitation across elements steers the beam so that a small transducer sweeps a large area
Array Scanheads
Arrays

- Ultrasound waves from different elements sum
- Adjust **timing** of excitation across the elements to steer and focus the beam
Electronic arrays control the excitation time of multiple transducer elements to steer and focus the ultrasound beam.

Steered

Focused

Steered and Focused

Sector scan
Resolution

- **Detail** (geometric) resolution
- **Temporal** (frame-rate) resolution
Arrays

- **Transducer elements in linear electronic arrays are not symmetric**
  - Beam pattern is not symmetric
Detail Resolution

- Axial: along the scan line (depth)
  - Axial Resolution = $1/2 \times$ (Spatial Pulse Length)
  - Constant with depth
  - Improves with increased frequency

- Lateral: perpendicular to the scan line within the image plane
  - Lateral Resolution = Beam width
  - Varies with depth
  - Improves with focusing and with increased frequency
# Axial Resolution

## Axial Resolution in B-Scans

<table>
<thead>
<tr>
<th>Transducer</th>
<th>Axial Distance</th>
<th>Pulse</th>
<th>Resolved Echoes</th>
<th>Not Resolved</th>
</tr>
</thead>
</table>

- **Transducer**
- **Axial Distance**
- **Pulse**
- **Resolved Echoes**
- **Not Resolved**
Axial Resolution

- **Axial Resolution** = (Spatial Pulse Length) / 2

- **SPL** = (# cycles/pulse) x \( \lambda \)

- **Improve axial resolution by**
  - reduced number of cycles
  - increased frequency

\[ \lambda = \frac{c}{f} \]

- Wavelength is affected by frequency and the medium
- Ultrasound wavelengths in tissue are less than 1 mm

\( c \) = propagation speed
\( f \) = transmit frequency
Axial Resolution

Example: 3-cycle pulse

<table>
<thead>
<tr>
<th>Transducer Type</th>
<th>Formula</th>
<th>Calculation</th>
<th>Axial Resolution</th>
</tr>
</thead>
<tbody>
<tr>
<td>5 MHz transducer</td>
<td>( \lambda = \frac{c}{f} )</td>
<td>( \frac{1540 \text{ m/s}}{5 \times 10^6 \text{ Hz}} )</td>
<td>0.308 mm</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>( \frac{3 \times \lambda}{2} )</td>
</tr>
<tr>
<td>10 MHz transducer</td>
<td>( \lambda = \frac{c}{f} )</td>
<td>( \frac{1540 \text{ m/s}}{10 \times 10^6 \text{ Hz}} )</td>
<td>0.154 mm</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>( \frac{3 \times \lambda}{2} )</td>
</tr>
</tbody>
</table>
Wavelengths

Spatial Pulse Length (SPL)

Boundary separation

1

resolved

> ½ SPL

1

unresolved

< ½ SPL

Higher frequency

resolved

> ½ SPL

UW Radiology
Lateral Resolution

LATERAL RESOLUTION IN B-SCANS

AXIAL DIRECTION

LATERAL SCANNING DIRECTION

BEAM WIDTH

RESOLVED STRUCTURES

NOT RESOLVED

d

R

NOT RESOLVED

NOT RESOLVED
Phased array transducer

Lateral resolution varies with depth

Beam diameter

Point targets

Lateral resolution

Image of point targets
Axial and Lateral Resolution
Axial and Lateral Resolution

Plot of image brightness from a string target

Imaging depth: 4.8 cm
Target depth: 2.3 cm
Focal depth: 3.5 cm

Beam direction

ATL L10-5 on HDI 3000

Leotta 1998
# Detail Resolution

<table>
<thead>
<tr>
<th>MHz</th>
<th>Axial resolution</th>
<th>Lateral resolution</th>
<th>Wave length (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>3.0</td>
<td>1.1 mm</td>
<td>2.8 mm</td>
<td>0.5</td>
</tr>
<tr>
<td>4.0</td>
<td>0.8 mm</td>
<td>1.5 mm</td>
<td>0.375</td>
</tr>
<tr>
<td>5.0</td>
<td>0.6 mm</td>
<td>1.2 mm</td>
<td>0.3</td>
</tr>
<tr>
<td>7.5</td>
<td>0.4 mm</td>
<td>1.0 mm</td>
<td>0.2</td>
</tr>
<tr>
<td>10.0</td>
<td>0.3 mm</td>
<td>1.0 mm</td>
<td>0.15</td>
</tr>
</tbody>
</table>

**Tradeoff:**
- Increased frequency ➔ Improved resolution
- Increased frequency ➔ Increased attenuation
Frequency Tradeoffs

Increased frequency ➔ Improved resolution
Increased frequency ➔ Increased attenuation

- Use lower frequencies for deeper structures
- Use highest frequency that can penetrate to the depth of interest

### Transcranial
1.5 - 2.0 MHz

### Cardiac
2.0 - 5.0 MHz

### Abdominal
2.0 - 5.0 MHz

### Musculoskeletal
5.0 - 12.0 MHz

### Peripheral Vascular
7.0 - 15.0 MHz
Beam Thickness

- Beam pattern perpendicular to 2D image plane
- Beam thickness (elevation) generally larger than lateral beam width
- Fixed focus set by acoustic lens
Beam Thickness

A Resolution components in 3-D space.

B Elevational Profile of Ultrasound Beam with depth

Acoustic Lens

Lateral

Slice thickness (elevational)

Axial
Array Resolution

Beam from an array is not symmetric

→ ‘anisotropic’ voxels

→ axial < lateral < elevation

Beam pattern varies with depth
Resolution

- **Detail** (geometric) resolution
- **Temporal** (frame-rate) resolution
Temporal Resolution

**Frame Rate:** the number of 2D images that can be produced per second

- Decreases with increasing imaging depth
- Decreases with increasing number of scan lines
Temporal Resolution

**Pulse Repetition Frequency**

- If echoes can arrive from as far as depth R, then we must wait at least until $t = 2R/c$ to transmit the next pulse
- Therefore the Pulse Repetition Frequency (PRF) must be $\leq c/2R$

**Frame Rate**

- Each pulse echo is used to construct one line of a 2D B-mode image
- If the B-mode image is made up of $n$ lines, then the time to scan one 2D frame is $n(2R/c) = n/PRF$
- Therefore the maximum Frame Rate is

$$FR_{\text{max}} = PRF/n = c/2Rn$$

$R = \text{range} = \text{maximum depth}$
$n = \text{number of scan lines per frame}$
Frame Rate

Example

- 1 pulse per scan line
- 8-cm imaging depth (R)
- 128 scan lines per frame (n)

→ maximum PRF = c/2R = 9625 Hz
→ maximum Frame Rate = PRF/n = 75 frames/sec
Doppler Ultrasound
Doppler Imaging

Stationary transducer

- Frequency unchanged

Stationary reflector

Reflector moving towards transducer
- Frequency increased

Reflector moving away from transducer
- Frequency decreased
The Doppler Shift

- The Doppler shift is the change in the frequency of sound due to motion of the source of the sound or the observer (or both).

- It equals 2 times the transmit frequency multiplied by the velocity and the cosine of the angle of incidence, all divided by the propagation speed of sound in human soft tissue.

\[ \Delta f = \frac{2 \cdot v \cdot f_t \cdot \cos \theta}{c} \]

- \( v \) = velocity
- \( f_t \) = transmit frequency
- \( \theta \) = angle of insonation
- \( c \) = speed of sound in human soft tissue
Doppler Imaging

- Visualization of anatomy and blood flow
  - Frequency shifts due to moving scatterers (red blood cells) are measured and displayed

- Color Doppler
  - 2D image showing presence, speed, direction, and character of blood flow

- Spectral Doppler
  - Detailed flow measurement at a single location
Basic Doppler Instrument

Continuous Wave

Transmitter amplifier

Receiver amplifier

Oscillator ($f_0$)

Demodulator

High pass filter

Doppler shift analyser

Monitor

Transducers

$\ f_0 = $ transmit frequency

$\ f_e = $ echo frequency

$\ f_D = $ Doppler frequency

Adapted from Amersham Medical
Detect changes in frequency of the reflected signal

Transmit frequency is used as a reference

\[ f_0 = \text{transmit frequency} \]
\[ f_e = \text{echo frequency} \]
\[ f_D = \text{Doppler frequency} \]
Demodulation

\( f_0 = \text{transmit frequency} \)
\( f_e = \text{echo frequency (received)} \)
\( f_D = \text{Doppler frequency} \)
Demodulation

Input wave

transmit

receive

Output wave

multiply

Mixer

Low-pass filter

Doppler shift
Doppler Imaging

- Visualization of anatomy and blood flow
  - Frequency shifts due to moving scatterers (red blood cells) are measured and displayed

- Color Doppler
  - 2D image showing presence, speed, direction, and character of blood flow

- Spectral Doppler
  - Detailed flow measurement at a single location
Doppler Angle

- Angle between sound travel & flow
- 0 degrees
  - flow in direction of sound travel
- 90 degrees
  - flow perpendicular to sound travel
  - no Doppler shift

\[ \Delta f = \frac{2 v f_i \cos \theta}{c} \]
\[ \Delta f = \frac{2 v f_t \cos \theta}{c} \]

Doppler Angle = 0
Doppler Angle

- Flow vector can be separated into two vectors

\[ \Delta f = \frac{2 v f_t \cos \theta}{c} \]

Doppler Angle > 0

Flow parallel to beam

Flow perpendicular to beam
Doppler Angle

- Flow vector can be separated into two vectors

\[ \Delta f = \frac{2 \, v \, f_t \cos \theta}{c} \]

Instrument can only measure the flow parallel to the beam

Doppler Angle > 0
Doppler Angle

- Flow vector can be separated into two vectors

\[ \Delta f = \frac{2 v f_t \cos \theta}{c} \]

\[ v = \frac{c \Delta f}{2 ft \cos \theta} \]
Color Doppler

Color flow systems represent the velocity in each pixel as a single value represented by a color selected from a color scale.

Note: velocity estimate changes with angle
Pulse wave systems measure the Doppler frequency spectrum at a specified depth and display the calculated velocities as a function of time.
Color Doppler guidance to sites of interest for Spectral Doppler
Three-Dimensional Ultrasound Imaging
3D Ultrasound

• Limitations of 2D ultrasound
  • 2D slices through a 3D structure
  • spatial relationships (between images and studies) are not preserved

• Benefits of 3D ultrasound
  • robust displays enhance interpretation
  • measurements require fewer geometric assumptions

• Acquisition methods
  • relate multiple 2D images in a 3D coordinate system
  • capture data in a 3D volume
3D Ultrasound Methods: Mechanical

Medison

GE

Mechanical scans
3D Ultrasound Methods: Freehand

Freehand systems

Magnetic Ascension Technology

Optical Image Guided Technologies

Articulated arm FARO
3D Ultrasound Methods: Volume Scan

2D array transducer

- N x N arrays are used to steer the ultrasound beam in both the azimuth and elevation directions
- Interrogate a pyramidal-shaped region and produce a volumetric image at high speeds without moving the transducer
- Recently-developed transducers include a 64 x 64 = 4096 element array operating at 3.5 MHz

Stennet, von Ramm
Carnegie Mellon / Duke
**3D Ultrasound Methods: Volume Scan**

**2D array transducer**

- N x N arrays are used to steer the ultrasound beam in both the azimuth and elevation directions.

- Interrogate a pyramidal-shaped region and produce a volumetric image at high speeds without moving the transducer.

- Recently-developed transducers include a 64 x 64 = 4096 element array operating at 3.5 MHz.

Images showing volume scans of various regions, including Aortic / Tricuspid / Mitral and Tricuspid / Mitral.
A standard ultrasound system is modified to relate multiple 2D images in a 3D reference coordinate system.
3D Ultrasound: Scanhead Tracking

- Relate pixel locations in ultrasound image with points in the 3D reference coordinate system
  - R: tracking output
  - S: calibration
  - I: in-plane pixel location
  - P: 3D pixel location
Volume Reconstruction

- Insert 2D image data into a regular 3D grid
- No manual interaction required
- Variety of display options

Image acquisition → Image stack → 3D volume
Image-to-Volume Processing

- Calculate volume bounds in transmitter coordinate system based on all images
- Specify voxel size in mm
- Calculate 3D location of pixel
- Calculate voxel number corresponding to pixel’s 3D location
- Insert pixel value in voxel

Coordinate systems

\( (x,y,z) \): magnetic transmitter

\( (r,c) \): image

\( (i,j,k) \): reconstructed volume
Shoulder Rotator Cuff

Images acquired from multiple windows

3D volume reconstruction
Rotator Cuff Thickness

- Tendon thickness measurements show changes in morphology in a subject with an acute tear of the left supraspinatus tendon. In particular note
  - the nearly uniform thickness of the normal tendon ($\approx 7$ mm)
  - the absence of the tendon on the anterior side of the bone in the injured shoulder
  - a pronounced bulge of the retracted end of the torn tendon (thickness $\approx 14$ mm)

Leotta and Martin
UMB, 26:509, 2000
Surface Reconstruction

- The vessel is traced on each image
  - manual or semi-automated
- Outline points are connected to create a surface for visualization and measurement
Surface Reconstruction

- Image outlining for structure measurement
- Segment multiple structures of interest
- 3D viewing window helps assess/guide tracing

Cardiovascular Research Training Center, UW
Surface Reconstruction

Left Ventricle: Endocardium

Traced Borders
Fitted Mesh: 500 faces
Fitted Mesh: 2000 faces
Final Surface

Leotta et al.
J Am Soc Echocardi, 1997
Cardiac Shape

- Compare individual reconstructed surfaces to a normal model derived from imaging of a representative population.

Munt et al.
J Am Soc Echocard, 1998
Anterior view of an aneurysm repaired by placement of an endovascular graft

Leotta et al., J Vasc Surg., 2001
Repaired aneurysm imaged 2 weeks (left), 6 months (center) and 1 year (right) after graft placement
Computational Flow Modeling for Dialysis Access Surgical Planning

3D surface model

Dialysis Access Arteriovenous Fistula
Serial Study: Vein Graft Revision

- Vein graft with progressing stenosis at site proximal to a PTFE patch angioplasty repair
  - Femoral to above-knee popliteal reversed saphenous vein graft
- Original graft: November 1996
- Revision: November 2003

6 months post-revision
Serial Study: Vein Graft Revision

PTFE Interposition Graft

PTFE Patch Angioplasty

Stenosis

6 months post-revision
Serial Study: Vein Graft Revision

1 month

- Vein graft with progressing stenosis at site proximal to a PTFE patch angioplasty repair
  - Femoral to above-knee popliteal reversed saphenous vein graft
  - Revision: November 2003

6 months

16 months
Serial Study: Vein Graft Revision

6 months

16 months

Axial Velocity over the Cardiac Cycle

- Turbulent jet impinging on vessel wall leads to dilation over time

McGah et al., J Biomech Eng, 2011
Serial Study: Vein Graft Revision

Wall Shear Stress: stress (force per unit area) that is applied parallel or tangential to a face of a material

McGah et al., J Biomech Eng, 2011